Properties of a new high-ratio anti-scatter grid in digital mammography

Poster No.: C-2647
Congress: ECR 2018
Type: Scientific Exhibit
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Keywords: Breast, Radiation physics, Mammography, Instrumentation
DOI: 10.1594/ecr2018/C-2647

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Aims and objectives

The performances of a new high-ratio DBT-compatible A/S grid were evaluated.

When introduced in mammography in 1978, the goal of an anti-scatter (A/S) grid was to increase image contrast. However, the reduced contribution of scattered radiation to the image receptor entrance air kerma (IREAK) had to be compensated by an increase in primary radiation and patient dose to maintain the IREAK value needed by the analog image receptors (screen/film combinations) to provide optimal images. Since that, A/S grids have been perceived as improving image quality at the expense of an increased dose to patients. When film/screens were replaced with digital image receptors, image display became independent from image acquisition, and it has been possible to restore image contrast by processing [1-2]. It could therefore be thought that a grid is not useful anymore. This is true for the deterministic degradation of image contrast, but the quantum noise of scattered photons irreversibly degrades the contrast-to-noise ratio of images. For this reason, the presence of an A/S grid can be fully justified if the reduction in scatter noise by the grid more than compensates the attenuation of non-scattered photons; new metrics have been proposed to describe this benefit [3-5]. Evaluations of the net balance of the presence of a grid in digital mammography have been done [5, 7, 8]. This is quantified by a "new quality parameter: the Image Improvement Factor or Q-factor (Q) which better describes the properties of the anti-scatter grid, especially for digital detector" [6], introduced with the 2013 revision of the IEC 60627 standard. "This factor better describes the properties of anti-scatter grids than the Grid Exposure Factor B and the Contrast Improvement Factor K, especially for digital detector applications. Namely, [for a constant entrance dose] the signal-to-noise ratio (SNR) for digital X-ray detectors is increased proportionally with the square root of the factor Q when an anti-scatter grid is applied. » [6]. Q is equal to the square of $K_{SNR}$ as defined in [3].

As a consequence, the presence of an A/S grid is beneficial as soon as Q > 1.

In addition to regular 2D mammography, using A/S grids for digital breast tomosynthesis (DBT) has been evaluated [9-12].

This work intends to measure the Image Improvement Factor of a new high-ratio DBT-compatible A/S grid and compare it to the values of Q provided by the A/S grid previously used in the same application.
Methods and materials

The previous grid (Grid A) and the new grid (Grid B) are linear focused grids, with a focusing distance of 600 mm. Both are DBT-compatible, and therefore their 15 µm lead strips are disposed parallel to the chest wall side of the breast-support, with the central line at the chest wall edge of the grid. They are protected by carbon-composite covers on both sides. The two grids are used in DBT equipment, SenoClaire for Grid A and Senographe Pristina™ for grid B (both from GE Healthcare, Chicago, IL, USA).

Grid A has a 5:1 ratio, 104 strips/cm, with graphite interspaces. Grid B has a 11:1 ratio, 67 strips/cm and fibre interspaces. Both are used in reciprocating mode, with amplitudes of 400 µm (grid A) and 2 mm (grid B).

The Image Improvement Factors (Q) of each grid were measured. Instead of a direct measurement of CNR with and without the grids as in [5], Q was "evaluated as the ratio of the square of the Transmission of Primary Radiation (T_p) to the Transmission of Total Radiation (T_t)" [6].

\[ T_t = 1/B \]

\[ K = T_p/T_t \]

\[ Q = T_p^2/T_t = T_p^2.B = K.T_p \]

However, in deviation with the standard, both grids were measured in regular operating conditions in their respective equipment, using clinical beam qualities and the image receptor as the detector, in reciprocating mode, with the carbon composite breast support in place.

Both systems use an X-ray tube with a rhodium anode. For grid A a 25 µm rhodium filter was used at 30 kV and for grid B a 30 µm silver filter at 34 kV. This difference has not been considered as significant knowing that the influence of beam quality on scatter to primary ratio has been demonstrated as negligible [13].

The primary (T_p) and total (T_t) transmissions were measured using PMMA as scattering material, for thicknesses t= 0 to 70 mm.
**Measuring Primary transmissions**

The primary transmission $T_p$ (transmission of non-scatter radiation) was measured in narrow-beam, low-scatter conditions. PMMA slabs were placed directly at the output of the tube head, between two radiopaque steel plates with aligned holes. The steel plates are assembled to form a support for the PMMA slabs (Fig. 1 on page 5 and Fig. 2 on page 5); the top plate was designed to be inserted in place of the face-shield, using the the same rails.

Images were acquired with a current-time product allowing to reach image levels equal to or higher than typical used clinically.

Image levels were measured in small regions of interest (ROIs) included in the image of the collimating holes, with the grid present, then removed (Fig. 3 on page 6). For each PMMA thickness, $T_p$ was computed as the ratio of image levels respectively with and without the grid.

The operation was repeated for the two grids.

**Measuring Total transmissions**

The total transmission $T_t$, (total transmission by the grid of primary and scatter) was measured in broad-beam, high scatter conditions. PMMA slabs of the same thicknesses as for narrow-beam conditions and larger than the full fields of view were placed on the breast support, covering the full detector field. The regular system collimator was wide open (23x31 cm for grid A, 23x29 cm for grid B).

Image levels were measured in the same ROIs as for $T_p$ measurements with the grid present, then removed (Fig. 4 on page 7). For each PMMA thickness, $T_t$ was computed as the ratio of image levels respectively with and without the grid.

**Computing Image Improvement factors**

For each PMMA thickness, $Q$ was computed as the ratio of $T_p^2$ and $T_t$. 


Fig. 1: PMMA support for narrow-beam low-scatter conditions; viewed from top left.

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Fig. 2: PMMA support for narrow-beam low-scatter conditions; viewed from front and bottom.

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Fig. 3: Primary transmission measuring set up. a) with grid in; b) with grid out

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Fig. 4: Total transmission measuring set up. a) with grid in; b) with grid out

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Results

For primary transmission, the narrow-beam collimation tool was used, with the different heights of PMMA plates; measurements were done in the front central hole (Fig. 6 on page 11).

Results of measurements and computations of $T_p$, $T_t$, and $Q$ are provided in Table 1 on page 12 for grid A and Table 2 on page 13 for grid B.

For grid A, $T_p$ values were found to be between 0.69 (t=0) and 0.72 (t=70 mm).

For grid B, $T_p$ values were found to be between 0.675 (t=0) and 0.71 (t=70 mm).

The primary transmission of grid B is therefore lower than that of grid A, from 2.3% in absence of object to 1.5% at maximum thickness (70 mm PMMA). The variation with thickness can be explained by some beam hardening.

For grid A, the values of "bucky factor" B were found between 1.47 (t=0) and 2.28 (t=70 mm).

For grid B, the values of "bucky factor" B were found between 1.51 (t=0) and 2.65 (t=70 mm).

In both cases the variation of B with thickness was linear.(Fig. 8 on page 15)

The Grid Exposure Factor B of grid B is increased by 2.7% (t=0) to 14% (t=70%) relative to grid A. In film/screen imaging the exposure to the image receptor should be increased in that proportion, with the benefit of a contrast improvement factor increase by 0.4% (t=0) to 12.7% (t=70%).

Of more interest is the Image Improvement Factor Q, with the following results (Fig. 9 on page 16)

- For grid A: from 0.70 (t=0) to 1.18 (70 mm); $Q \geq 1$ for $t \geq 42$ mm
- For grid B: from 0.69 (t=0) to 1.34 (70 mm); $Q \geq 1$ for $t \geq 34$ mm
Fig. 4: Total transmission measuring set up. a)with grid in; b)with grid out

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Fig. 5: PMMA plates support for narrow-beam collimation

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Fig. 6: Image obtained in narrow-beam conditions. The dimensions are stated in the image receptor plane. Image levels are measured in the circular ROI on-axis, chestwall side.

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<table>
<thead>
<tr>
<th>PMMA Thickness (mm)</th>
<th>$T_p$ measurements</th>
<th>$T_t$ measurements</th>
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<tr>
<td></td>
<td>With grid</td>
<td>No grid</td>
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<tr>
<td>0</td>
<td>1448.3</td>
<td>2096.1</td>
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<tr>
<td>70</td>
<td>1363.5</td>
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**Table 1**: Measurement results and computations of $T_p$, $T_t$, and $Q$ for grid A

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<table>
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<tr>
<th>PMMA Thickness (mm)</th>
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**Table 2:** Measurement results and computations of Tp, Tt, and Q for grid B

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Fig. 7: Variations of Primary Transmission (Tp) with PMMA thickness for grids A and B

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**Fig. 8:** Variations of Grid Exposure Factor ("Bucky factor") $B$ with PMMA thickness for grids A and B

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Fig. 9: Variations of Image Improvement Factor Q with PMMA thickness for grids A and B

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Conclusion

The reduction in PMMA thickness were Q # 1 (t= 42 mm PMMA for grid A, t= 34 mm PMMA for grid B) demonstrates a significant progress from grid A to grid B. These values should be compared to the generally accepted average-breast equivalent of 45 mm PMMA [15], even if a direct comparison between PMMA thickness and breast thickness is not fully valid here since usual equivalences are based on attenuation, not scatter. In comparison, a threshold of 65 mm was found in [5]. With a different method (CDMAM contrast/detail phantom) and a first-generation digital mammography equipment (Senographe 2000D, GE Healthcare) equipped with the original 5:1 fibre interspaced grid, "For 5 and 7 cm [PMMA], no significant difference was found" [14] between images acquired with and without a grid. Considering the distribution of PMMA-equivalent thicknesses [5], only a small fraction of the population would then benefit from the presence of a grid with these values. With grid A, the beneficial thickness is close to that of the average breast, meaning that approximately half the population examined benefits from the presence of this grid (assuming a normal distribution as an approximation). Since the patient dose is increasing with breast thickness, the population dose reduction is more significant. With grid B and a threshold reduced to 34 mm, the proportion of a typical western population which can benefit from the presence of the grid [17] is raised to more than 90%.

Consequently, provided the automatic exposure control design is not intended to maintain a constant image receptor entrance air-kerma, using grid B allows maintaining a given CNR with a further moderation of the patient dose for thick breasts. Thanks to the DBT-compatible disposal of this grid, this benefit applies to both 2D and 3D mammography with no distinction.
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